A novel deformable lung phantom with programably variable external and internal correlation

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Purpose: Lung motion phantoms used to validate radiotherapy motion management strategies have fairly simplistic designs that do not adequately capture complex phenomena observed in human respiration such as external and internal deformation, variable hysteresis and variable correlation between different parts of the thoracic anatomy. These limitations make reliable evaluation of sophisticated motion management techniques quite challenging. In this work, we present the design and implementation of a programmable, externally and internally deformable lung motion phantom that allows for a reproducible change in external–internal and internal–internal correlation of embedded markers.

Methods: An in-house–designed lung module, made from natural latex foam was inserted inside the outer shell of a commercially available lung phantom (RSD, Long Beach, CA, USA). Radiopaque markers were placed on the external surface and embedded into the lung module. Two independently programmable high-precision linear motion actuators were used to generate primarily anterior–posterior (AP) and primarily superior–inferior (SI) motion in a reproducible fashion in order to enable (a) variable correlation between the displacement of interior volume and the exterior surface, (b) independent changes in the amplitude of the AP and SI motions, and (c) variable hysteresis. The ability of the phantom to produce complex and variable motion accurately and reproducibly was evaluated by programming the two actuators with mathematical and patient-recorded lung tumor motion traces, and recording the trajectories of various markers using kV fluoroscopy. As an example application, the phantom was used to evaluate the performance of lung motion models constructed from kV fluoroscopy and 4DCT images.

Results: The phantom exhibited a high degree of reproducibility and marker motion ranges were reproducible to within 0.5 mm. Variable correlation was observed between the displacements of internal–internal and internal–external markers. The SI and AP components of motion of a specific marker had a correlation parameter that varied from -11 to 17 . Monitoring a region of interest on the phantom's surface to estimate internal marker motion led to considerably lower uncertainties than when a single point was monitored.

Conclusions: We successfully designed and implemented a programmable, externally and internally deformable lung motion phantom that allows for a reproducible change in external–internal and internal–internal correlation of embedded markers. © 2019 American Association of Physicists in Medicine [https://doi.org/10.1002/mp.13507]

Key words: deformable lung phantom, programmable, respiratory motion management, surrogate-based lung motion model, variable correlation

1. INTRODUCTION

In thoracic and abdominal radiation therapy (RT), respiration-induced anatomic motion, if not accounted for, can limit the geometric and dosimetric accuracy of the delivered dose.¹ Improving the accuracy and precision of RT dose delivery requires the ability to accurately measure the position of the tumor and the surrounding organs at risk (OARs) throughout the treatment duration^{2,3} and to estimate the effect of the motion on the delivered dose to the tumor target and OARs.^{4–7}

To date, the most common approach for addressing respiratory motion has been to assume that the breathing pattern is periodic as a function of time.^{4,5} However, it is well known that under free-breathing conditions, many patients have irregular breathing patterns such that a periodic function is an inadequate characterization of respiratory motion.^{2,6–11} In light of this fact,

researchers have resorted to more sophisticated models in the framework of motion prediction¹² and real-time estimation of respiratory motion models based on local regression methods.^{13–18} A clinical example of implementation of such a model is the Cyberknife robotic radiosurgery system (Accuray, Inc.; Sunnyvale, CA, USA), which uses real-time fluoroscopic tracking to build and update the relationship between external skin markers and internal fiducials placed inside or near the tumor.¹⁹ Irrespective of whether the markers are internal or external, two main assumptions underlie current surrogate-based tumor tracking: (a) the motion of the external or internal surrogates is correlated with that of the tumor target and (b) this correlation, which is estimated at the initial simulation (SIM-CT), remains constant cycle-to-cycle and fraction-to-fraction. For example, if we estimate from the SIM-CT that 2 mm AP motion of an external surrogate corresponds to 2 mm AP, 3 mm SI, and

1.5 mm left–right motion of the tumor centroid, it is unlikely that the absolute values will remain unchanged throughout the duration of the fraction, much less, fraction to fraction, because of respiratory cycle-to-cycle variations in proportion of engagement of abdominal vs thoracic muscles.

Empirical validation of these models is usually performed using programmable motion phantoms that attempt to mimic, as closely as possible, anatomic changes during respiration. These phantoms can evaluate and test a given motion management technique before it is deployed for clinical use. We can categorize these phantoms as rigid²⁰ and deformable.^{21–29} An example of a fully rigid (rigid interior–rigid exterior) phantom was provided by Starkschall *et al.*, who modified a commercially available rigid motion phantom (Respiratory Position Management [RPM]; Varian Medical Systems; Palo Alto, CA, USA) to evaluate a set of parameters useful for assessment of the quality of four-dimensional (4D) computed tomography (CT).²⁰ A slightly irregularly shaped wheel, with the RPM placed on top of the phantom, allowed vertical displacement, a common indicator of anterior–posterior (AP) motion of the chest during respiration.

In deformable phantoms, the extent of deformability can dictate the potential of the phantom to resemble the human anatomy during respiration. Internal deformability is assumed to replicate the internal target/anatomy motion, and external deformability attempts to reproduce the visible motion of the chest and abdomen during respiration (external surrogate). Some extent of internal deformability is included in the majority of existing phantoms, whereas only a few produce the external motion. For example, Serban *et al.* used a cylinder filled with water and a latex balloon stuffed with dampened sponges to make an externally rigid–internally deformable (rigid + deformable) phantom.²¹ The internal motion was created using a piston attached to a programmable motor. The same phantom was later modified to simulate respiratory-induced lung tumor motion and deformation while the delivered dose was measured at different positions in the lung and the tumor in real time.²⁴ Chang *et al.* reported on a similar phantom with a more complex approach for transferring periodic motion to the deformable interior, including a silicon membrane instead of a perforated lid.²³ Nioutsikou *et al.* delivered and measured the dose in a cylindrical tumor that was embedded inside a tissue-equivalent breathing rigid + deformable phantom.²⁵ Kashani *et al.* designed a phantom by inserting foam infused with iodine inside the exterior of a rigid commercial thoracic phantom (RS-330; Radiology Support Devices; Long Beach, CA, USA). The deformable interior was then compressed and decompressed to simulate the SI motion of respiration.²⁷ Niu *et al.* proposed a method to build an internally rigid + deformable phantom and used gel dosimeters to experimentally validate deformable dose algorithms by measuring their 3D distribution during periodic motions and compared the results with delivered dose to the static control.²⁶ Remmert *et al.* used porcine heart–lung explants to build a rigid + deformable lung phantom with periodic motion to evaluate a new 4D-MR imaging method. The need for sorting the

images was fulfilled by positioning of the piston rod and monitoring the motion of the dynamic lung phantom, because the exterior of the phantom was rigid. In a clinical setup, AP motion of the chest is the common external surrogate used for such cases.²⁸

Court *et al.* used a deformable + deformable breathing phantom (Radiology Support Devices, Inc.; Long Beach, CA, USA) to compare the delivered dose and CT numbers for different treatment planning methods for tumors that were made of flexible resin material.²² Two independent motion actuators were used: a pump that inflated and deflated the lung and a motor that moved the “tumor” inside the lung based on patient-derived trajectories. However, the two actuators, and therefore their corresponding generated motions, were independent of each other.

With the exception of the phantom described by Court *et al.*, thoracic phantoms which are described in the literature as “deformable” are typically rigid on the exterior surface and deformable within the interior cavity. Such designs do not adequately represent respiration because, during the respiratory cycle, the tumor and the surrounding critical organs move relative to one another in the human body. We have previously presented an externally and internally deformable phantom that used a high-precision motion platform (MP) programmed with patient-derived motion trajectories, to actuate motion by compressing and decompressing a deformable insert.²⁹ In this phantom, the correlated internal and external motion of the human anatomy was replicated. However, as a consequence of having the same actuator generating motion, the correlation between the motion of the external and internal anatomy was constant. Such constant correlation is unrealistic because, as a result of varying amounts of engagement of the thoracic and abdominal muscles during human respiration, the internal–external and internal–internal correlations vary from one respiratory cycle to another.^{5,30} This limitation of a constant internal and external correlation is present in all phantom designs that utilize a single motion actuator.

In this work, we present the design and implementation of an anthropomorphic lung motion phantom that is externally and internally deformable and can be programmed with mathematical (e.g., sinusoidal) or patient-derived motion trajectories so as to simulate respiratory motion. The ability of the phantom to programmably produce complex and irregular motion is demonstrated by imaging external and internal markers using kV fluoroscopy. A potential application of the phantom is described, where we compare the performance of different motion models constructed from kV fluoroscopic images, or from 4DCT to estimate marker trajectories over multiple breathing cycles.

2. MATERIALS AND METHODS

2.A. Phantom design

In order to closely mimic respiration-induced motion and deformation in the human thoracic anatomy, the phantom was designed with the following characteristics:

1. Variable correlation between external surface displacement and internal marker motion;
2. Two independent actuators of motion to represent the two major groups of muscles that actuate human respiratory motion: (a) thoracic–intercostal; and (b) abdominal;
3. Programmable (and therefore) reproducible deformation and the interplay between internal–internal and internal–external objects within the phantom; and
4. Cycle-to-cycle variations in amplitude, frequency, baseline position, and hysteresis.

The phantom we present in this study is the assembly of two major parts: an anthropomorphic exterior and an MP.

2.A.1. Anthropomorphic exterior

A commercially available deformable lung phantom (RS-1500; Radiology Support Devices, Inc.; Long Beach, CA, USA) with a rigid posterior half and a flexible anterior half was used as the shell for anthropomorphic exterior of the phantom. An in-house–designed natural latex foam embedded with radiopaque fiducial markers was inserted inside this shell.

2.A.2. Motion platform

The MP consisted of two independently programmable linear actuators (Washington University; St. Louis, MO, USA³¹). The actuators were linked to the anthropomorphic part via a set of T-slots including the SI and AP arms (Fig. 1). The SI arm directly moved the SI push plate to compress and decompress the foam in the SI direction. The AP arm was coupled to the AP bending strips via the AP push plate and

FIG. 1. Schematic 3D drawing of the phantom design. Two linear actuators were connected to the anthropomorphic part (outer shell) via anterior–posterior (AP) and superior–inferior (SI) arms. The SI arm was connected to the SI push plate (shown in purple) and directly compressed and decompressed the foam in SI direction; the AP arm was connected to the AP push plate (shown in green) and deformed the foam in the AP direction by pushing the bending strips. The bending strips were composed of 3 layers of plexiglass and were attached to the interior of the phantom shoulder. Lock plates secured the phantom to the couch. [Color figure can be viewed at wileyonlinelibrary.com]

was used to generate AP motion. Actuator motion range in superior direction, without the exertion of any force was 6 cm. With the foam placed inside the phantom, this range was reduced to 4 cm in the superior direction and further force would lift the phantom and disturb the setup. Also, during the setup, we had to ensure the foam did not lose contact with the push plate throughout the measurement since it would create plateaus in the markers' trajectories and could be mistaken with the plateaus caused by the phase-shift between the SI and AP components of input trajectory.

In a previous publication, we described the design and evaluation of a programmable lung motion phantom that was externally and internally deformable.²⁹ The primary motion of the markers in the phantom was in the SI direction, with a limited amount of marker AP motion (<3 mm). However, because the motion was generated by the same linear actuator, the phantom only allowed constantly correlated internal and external motion and failed to satisfy characteristics 1–3 of respiratory induced motion as described above. The current design used two independent actuators for SI and AP directions of motion which allowed the generation of variable correlation between internal and external motions.

2.A.3. Superior inferior motion actuator

Figure 1 displays a schematic of our phantom design. SI motion was primarily caused by pushing the inserts of the phantom in the craniocaudal direction via the SI push plate. The vertical location, the angle, and length of the connecting arm for the SI push plate were designed to be adjustable to enable easier experimental setup and teardown.

2.A.4. Anterior–posterior motion actuator

Two sets of 3-layer plexiglass strips (AP bending strips in Figs. 1 and 2) were positioned inside the shell using a multi-purpose cement. We used multiple layers of strips because they were found to permit an easier bend than a single thick strip, while, at the same time, were able to mechanically support deformation of the foam. The back-and-forth motion of the AP push plate bent the bending strips, which compressed and decompressed the foam in the AP direction. The position of point of maximum bend and the length for both bending strips were also designed to be adjustable. These factors were found to have a direct impact on the magnitude of AP motion in the foam. Additionally, the lengths of the SI and AP arms were also adjustable, which improved setup ease and reproducibility. At our typical setup position, we have measured that 1 cm displacement of the AP push plate results in approximately 0.9–1 cm of AP displacement in the bending strips.

Some properties of the trajectories programmable into the phantom are reported in Table I.

2.A.5. Embedded markers

A fluoroscopic projection of the interior of the phantom is shown in Fig. 2(a). A lateral topogram image acquired prior

to 4DCT measurements, and three views of the CT images of the phantom are shown in Fig 2(b) and 2(c). Markers were used in the imaging studies to investigate the characteristics of the motion and deformations generated in the phantom. Several spherical glass marbles (16-mm diameter) were planted in the foam along the phantom midline and are shown in Fig. 2 (m_{int5} , m_{int6} , m_{int7} , m_{int8} , m_{int9} , and m_{int10}). In addition, three tungsten balls (3-mm diameter) were placed on the surface of the phantom (m_{ext1} , m_{ext2} , and m_{ext3}). An additional metal tungsten ball (12-mm diameter) was attached to the end of the AP arm and used as a control marker for the AP actuator (m_{CTRL4}). Similarly, a part of the SI arm was used as control marker for the SI actuator (m_{CTRL11}).

2.A.6. Programmed trajectories

We considered two extreme cases of correlation between the SI and AP actuators by programming the actuators with in-sync and out-of-sync sinusoidal trajectories (Sin_{IN} and Sin_{OUT}). A phase shift of 0.6π was programmed for the Sin_{OUT} trajectory. The SI and AP amplitudes of these trajectories were 15 and 7.5 mm respectively.

Four 2D, patient-derived, tumor trajectories were also played in the MP. The trajectories were measured by CyberKnife Synchrony and reported by Suh et al.¹⁹ The data were inspected visually to exclude regions of possible tracking errors, including large transient tumor

TABLE I. The motion platform is capable of playing trajectories with characteristics listed here.

Properties of the trajectories programmable in the phantom		
	Minimum	Maximum
SI displacement (cm)	0	4
AP displacement (cm)	0	2
Respiratory rate (breath/minute)	1	50

AP, anterior–posterior; SI, superior–inferior.

position changes and long pauses. For each trajectory, we manually selected a continuous, artifact-free, 4-min interval. In order to match the MP required temporal frequency, the motion traces were resampled from 26.3 to 50.0 Hz using linear interpolation.

The four patient traces are presented in Fig. 3. These trajectories exhibited some of the common characteristics of lung tumor motion such as base-line shifts, cycle-to-cycle variations, and sudden jumps. For example, Fig. 3(a) shows a negative correlation between the AP and SI. Figure 3(b) shows a trajectory with limited AP motion and large cycle-to-cycle variations and a baseline shift. The SI and AP components shown in Fig. 3(c) exhibit a high correlation while the correlation between the these two components changes after approximately 110 s in Fig. 3(d). Additionally, a sudden

FIG. 2. Arrangement of the internal and external markers. (a) Tungsten balls, shown as Markers 1–3 (m_{ext1} – m_{ext3}), were placed on self-stick felt pads on top of the phantom. Sixteen-millimeter glass marbles were used as internal markers and were placed along the midline of the phantom. Six of these internal markers were segmented and tracked using intensity-based thresholding (m_{int5} – m_{int10}). Markers 4 (m_{CTRL4}) and 11 (m_{CTRL11}) were used as control markers for AP and SI actuators, respectively. AP bending strips were visible in the projection. These strips were bent by the AP push plate to compress and decompress the foam in AP direction. SI push plate directly compressed the foam to create SI motion in the phantom. (b) A lateral topogram/surview image of the phantom acquired prior to 4DCT acquisition. (c) Examples of the transverse, coronal, and sagittal slices of the phantom acquired from 4DCT. For each motion trace, internal markers in each of the 10 phases of 4DCT images were contoured in the RayStation TPS and marker centroid positions were recorded. [Color figure can be viewed at wileyonlinelibrary.com]

FIG. 3. 2D AP and SI patient-derived, tumor motion traces used in this study. (a) Traj1: SI and AP components were negatively correlated and a limited amount of AP displacement was observed (<5 mm) (b) Traj2: AP and SI components of motion were correlated and exhibited a base line shift. (c) Traj3: a direct correlation with comparable amplitude of motion was observed between the SI and AP components. (d) Traj 4: a change in correlation between the SI and AP was observed after 100 s. For all trajectories, a 5-s ramp was added at the start to avoid actuator lag and stall. Additionally, 40 s of deadtime was added to permit the operator leave the LINAC vault after starting the motion platform. [Color figure can be viewed at wileyonlinelibrary.com]

jump (such as a cough) was seen at $t = 250$ s for Traj3 [Fig. 3(b)]. Finally, all four traces exhibit cycle-to-cycle variations.

The linear actuators on the MP were able to move one full step (1.8°) in 20 ms, and sudden vertical jumps or large thrust loads (>70 lbs) resulted in motors stall and lag. To avoid these issues, a 5-s ramp connecting zero to the start of the trajectory was augmented. Additionally, 40 s of dead time was added to permit the operator of the MP to leave the LINAC vault before the start of the scan.

2.B. Study of phantom characteristics

2.B.1. Measurements

Motion characteristics of the phantom were studied using fluoroscopy according to the setup shown in Fig. 4(a). The lateral fluoroscopic images were acquired using the kV

imager of a Varian Truebeam LINAC at 80 kVp, 25 mAs, and rate of 7 fps. A couch mounted VisionRT (VRT) system was used to monitor the surface of the phantom.

2.B.2. Marker segmentation and tracking in Fluoro images

An intensity-based template-matching algorithm was used to track markers in each fluoro image. Tracking results were visually inspected frame-by-frame to ensure accurate tracking. A voxel-to-displacement magnification factor (mm/voxel) was obtained by dividing the segmented diameter of the marker as delineated in fluoro images by the physical dimension of the marker. The spatial dependence of the magnification factor for each marker was examined by using several fluoro images acquired at different instances of the phantom during the respiration cycle.

FIG. 4. Experimental setup of the phantom. (a) Phantom setup in the linac room with external markers that were placed on self-stick felt pads. Lock plates secured the phantom and the motion platform to the couch for imaging studies. Concurrent with the fluoroscopic acquisitions, the surface of the phantom was recorded at 15 fps using the couch-mounted VRT system. (b) Phantom setup on the CT couch. The varian respiratory position management was used as the breathing signal and 4DCT images with concurrent surface acquisitions were collected for each motion trace. [Color figure can be viewed at wileyonlinelibrary.com]

2.B.3. Motion reproducibility

In order to examine the reproducibility of the phantom, fluoroscopic measurements using Sin_{IN} , Sin_{OUT} , and Trajl were repeated on four separate occasions. Identical setup procedures and data processing methodology were used for all four measurements.

2.B.4. 4DCT measurements

The above fluoroscopic setup was repeated on the CT couch and phase binned 4DCT measurements (3 mm slice thickness, 0.44-s rotation time, pitch of 0.059, collimation = 16 mm $\times$ 1.5 mm) were performed using the varian real time position management (RPM) system as the respiratory surrogate. Simultaneous to the 4DCT acquisition, the VRT system was used to monitor the phantom surface as shown in Fig. 4(b). For each 4DCT acquisition, the RayStation treatment planning system (RaySearch Laboratories, Sweden) was used to contour the markers in each of the 10 phases of the 4DCT [Fig. 2(c)] and marker centroid positions were recorded for subsequent analysis.

2.C. Applications

2.C.1. Validation of the couch-mounted VRT system

By projecting a random pattern of speckles on the surface, the VRT system generates a point-cloud representation of that surface. However, the measurement process of this point cloud is prone to geometric distortions that could stem from factors such as room lighting conditions, presence of immobilizing devices and fixtures or chest hair. Due to its shallower angle, and closer proximity to the surface, the couch mounted system was more susceptible to such distortions. These distortions result in missing patches of data and must be handled with a surface reconstruction method. For this purpose, we implemented a signed surface distance (SSD) reconstruction method that takes as input the VRT point cloud and approximates the corresponding water-tight surface that is free of missing patches.³²

As an application of the phantom, we provide a validation of our surface acquisition and reconstruction methodology by comparing the VRT measured external marker displacements for $m_{\text{ext}1}$ – $m_{\text{ext}3}$, with their displacement measured using fluoroscopy. To do this, three 1 cm² $\times$ 1 cm² regions of interest (ROI) patches corresponding to the positions of $m_{\text{ext}1}$, $m_{\text{ext}2}$, and $m_{\text{ext}3}$ were selected. For each ROI, the estimated AP displacement using VRT + SSD was compared with the marker displacement measured from fluoroscopy.³²

2.C.2. Performance characterization of surrogate based motion models using kV fluoroscopy or 4DCT

Surrogate-based motion models use external surrogates such as abdominal height displacement to estimate the

corresponding internal anatomical state.^{15,16,18} We demonstrate a potential application of our phantom as a tool for the experimental verification and quality assurance of such motion models. To do this, we quantified the performance of different motion models that were generated based on images acquired from two imaging modalities: kV fluoroscopy or 4DCT. For all models, partial least square regression (PLS regression)³³ was used to relate the external surrogate(s) with internal marker displacement.

Using kV fluoroscopy, we compared the performance of three models that estimate the displacement of internal markers using information from the surface with different levels of complexity. The scalar model used the height of a single point ($m_{\text{ext}2}$) on the surface of the phantom as a surrogate. The height+velocity model used as surrogate both the displacement and the velocity of marker $m_{\text{ext}2}$. The numerical derivative was computed from the height using a four-point forward difference formula (accuracy = h^4).³⁴ Finally, the surface model used the deformation of a 20 cm² $\times$ 20 cm² ROI, discretized into 0.37 cm² $\times$ 0.37 cm² patches to estimate displacement of internal markers.

Partial least square regression was used to relate the surrogate(s) to the displacement of internal markers $m_{\text{int}5}$, $m_{\text{int}6}$, $m_{\text{int}7}$, $m_{\text{int}8}$, and $m_{\text{int}10}$. The first 10 s of the surrogate was used along with the actual marker displacements, measured from fluoro images, to train the models. The surrogates were then used to estimate marker displacement for the remaining 230 s.

For each motion trace T and marker N , the error $in_X^{T,N}$ (in mm) associated with model $X \in \{\text{scalar}, \text{height} + \text{velocity}, \text{surface}\}$ was defined as:

$$\epsilon_X^{T,N}(t) = S_X^{T,N} - S_{\text{fluoroscopy}}^{T,N} \quad (1)$$

where $S_X^{T,N}(t)$ is the marker N displacement for trajectory T at time t with model X , and $S_{\text{fluoroscopy}}^{T,N}(t)$ is the corresponding marker displacement measured from fluoro images.

To examine the feasibility of using VRT surfaces in conjunction with 4DCT for estimation of target positions over multiple cycles, a motion model based on 4DCT images was also considered. Due to the single-cycle, phase-averaged nature of the 4DCT measurement, and the continuous fashion by which VRT surfaces are acquired, a single-cycle, phase-averaged training set of VRT surfaces was needed to relate these two measurements. This training set was generated by tracking the position of a 2 cm² $\times$ 2 cm² patch near the RPM box in the VRT surfaces and applying a phase-sorting algorithm to assign phase indices ranging from 0% (end of inhale) to 50% (end of exhale) and back to 90% (end of inhale) to each acquired VRT surface. Using this phase assignment, 10 “phase-averaged” VRT surfaces were generated by averaging all VRT surfaces that belonged to the same phase. Once the training set was compiled, a PLS model (4DCT + VRT) was used to relate the phase-averaged VRT surfaces with the 4DCT measured marker positions. For each

observed VRT surface $\vec{S}(t)$, the PLS model output an estimate of marker centroid displacement.

The potential benefit of the VRT + 4DCT model over 4DCT alone was compared by using marker motions from fluoroscopy as ground-truth. For each trajectory, the single cycle 4DCT measurement was expanded over multiple cycles (4DCT expanded) using the phase-assignment described above. The error for marker N , at time t was computed using Eq. (1), where $X \in \{4DCT_{\text{expanded}}, 4DCT + VRT\}$ refers to the 4DCT measurement expanded over multiple cycles, or the marker displacement estimated using the 4DCT + VRT model.

3. RESULTS

3.A. Phantom characteristics

3.A.1. Motion reproducibility

Table II reports the average and standard deviation of marker motion range (in mm) for $m_{\text{int}5}$ in the SI and AP directions from four separate days of measurement. The standard deviation for this marker varied from 0.1 to 0.5 mm.

3.A.2. Deformable motion and variable correlation

Figure 5(a) shows the change in the correlation between the AP and SI displacement of $m_{\text{int}5}$, $m_{\text{int}7}$, $m_{\text{int}8}$, and $m_{\text{int}10}$ inside the phantom. Each respiratory cycle appears as a loop in the figure, and the presence of cycle to cycle variations resulted in a seen as different loops for each cycle. For each marker, its AP and SI displacements were found to depend on the input trajectories as well as the proximity of the marker to each of the AP bending strips and SI push plate. The correlation parameter (slope of the linear fit) between the SI and AP displacements varied from 2 to 11 (mm SI)/(mm AP) for $m_{\text{int}10}$ and $m_{\text{int}5}$, respectively with higher slopes associated with markers that moved further in the SI direction than AP direction. For markers closer to the AP bending strips (and hence further from the SI push plate), the slope of this fit was lower due to the dampening effect of the foam (lower SI motion) and higher magnitudes of AP motion.

The variation in the correlation between the AP and SI displacement of $m_{\text{int}5}$ with different trajectories is shown in Fig. 5(b). Common to all markers, the direction of the

correlation between the SI and AP motions was in agreement with the input trajectory. For $m_{\text{int}5}$, the correlation parameter between the marker's SI and AP displacements varied from +17 to -11. Additionally, presence of hysteresis in the marker's motion resulted in deviation from a linear relationship between marker's AP and SI components, with largest hysteresis observed for Sin_{OUT} and Traj4.

Correlation between the AP displacement of $m_{\text{ext}2}$ and the SI displacement of $m_{\text{int}6}$ for different trajectories is shown in Fig. 5(c). The slope of the linear fit between the internal SI and external AP directions was found to vary from -3.7 to -2.5, and was specific to the input motion trace. Similar to the Fig. 5(b), greatest hysteresis was observed for Sin_{OUT} and then Traj4. Other combinations of external-internal and internal-internal markers also exhibited similar behavior. The data from these are not shown for brevity.

3.A.3. Validation of surface monitoring and reconstruction

The presence of the external markers, $m_{\text{ext}1}$, $m_{\text{ext}2}$, and $m_{\text{ext}3}$, resulted in geometric distortions in the point cloud captured by the couch-mounted system and produced missing patches. The captured VRT point cloud, reconstructed using the SSD technique on a 100×100 grid and covering a $31 \text{ cm}^2 \times 31 \text{ cm}^2$ area, is presented in blue in Fig. 6. Prior to using the captured surface for further analysis, we investigated the accuracy of the VRT + SSD methodology for target position estimation using the ground truth fluoroscopy. The two measurements were in high agreements and sub-millimeter mean errors were observed. The error distributions were found to resemble a normal distribution. A box-plot of the error for each external marker and trajectory is presented in Fig. 6. For the majority of trajectories, the 95th percent confidence window ($2 \times \text{SD}$) was covered by a 1.2 mm window. The median error was found to be less than 0.5 mm with a 1 mm margin covering two standard deviations of the acquired points.

3.A.4. Performance characterization of surrogate-based motion models using kV fluoroscopy or 4DCT

Figure 7(a) shows a box plot comparison of estimated $m_{\text{int}5}$ displacement errors for each input trajectory from models based on kV images. All three models used information from the surface to estimate internal marker displacements. Largest errors were observed when the scalar model was used where the height of a single point was the input to the model. This model could not distinguish well between different inhale or exhale phases. Due to the linear nature of the PLS regression, at each instance in time, estimated marker displacements were averaged to generate a best fit linear curve. This led to large errors when non-linear hysteresis was observed. For example, the out of sync nature of the Sin_{OUT} trajectory resulted in an elliptic shaped path between the internal marker SI and the surface AP displacement, which could not be adequately captured

TABLE II. Motion range of $m_{\text{int}5}$ (in mm) average of four separate measurements for three trajectories.

	$m_{\text{int}5}$ motion range ($\pm \text{SD}$)	
	AP direction	SI direction
Sin_{IN}	1.4 (± 0.1)	18.0 (± 0.1)
Sin_{OUT}	2.4 (± 0.3)	9.4 (± 0.2)
Traj1	1.1 (± 0.2)	11.6 (± 0.5)

AP, anterior-posterior; SI, superior-inferior

FIG. 5. Correlation between internal and external displacements. (a) Observed correlation between the AP and SI displacement for internal markers, m_{int5} , m_{int7} , m_{int8} , and m_{int10} when Traj3 was played in the motion platform. The AP and SI components of this trajectory were highly correlated and moved in synchrony. The correlation (slope of the best-fit line) between marker motion in the AP and SI directions was different for each marker, and dependent on its position with respect to the AP bending strips and the SI push plate. (b) Observed correlation between marker m_{int5} AP and SI displacement for all input trajectories. Significant variation in slope, motion range and hysteresis were observed for different trajectories. (c) Correlation between AP displacement of m_{ext2} and SI displacement of m_{int6} . [Color figure can be viewed at wileyonlinelibrary.com]

FIG. 6. Captured VRT point cloud and the SSD reconstructed phantom surface. (a) Reconstructed surface (blue), with the initially acquired VRT point cloud (red). The point cloud has missing patches caused by geometric/optical distortions such as presence of wrinkle. Placement of the external markers, m_{ext1} , m_{ext2} , and m_{ext3} produced distortions and missing patches in the VRT captured point cloud. (b) VRT surface was reconstructed using the SSD algorithm. Positions of the external markers are highlighted in red. (c) Box-plot of the VRT error for marker displacement computed with respect to fluoroscopy. Trajectories are denoted on the x axis and include Sin_{IN} , Sin_{OUT} , and the four tumor traces, Traj1–Traj4. [Color figure can be viewed at wileyonlinelibrary.com]

using linear regression. As a result, an 8-mm window was required to cover the 95th error percentile. Large errors were also observed for Traj1 and Traj4. However, as shown by Fig. 7(b) and 7(c), the correlation between the AP and SI components of input trajectories changed mid-way for Traj1 [Fig. 7(b)] and Traj4 [Fig. 7(c)]. Consequently, the observed correlation that was used to train the model was no longer valid, and its use resulted in large disparities between the model and measurement. For example for Traj1 and Traj4, a 4 mm window was required to cover the 95th error percentile using the scalar model.

The height + velocity model integrated information regarding the direction and speed of motion, and the height to estimate marker displacement. As a result, the model

was more accurate than the scalar model, but was still limited when large disparities or a change in correlation between the SI and AP components of motion were encountered. In such situations, a single point was not an adequate representative of the internal displacement. For example, given the periodic nature of Sin_{OUT} , the model generated an accurate estimate of marker motion. However, errors for Traj1 and Traj4 were comparable to those from the scalar model, and a 4-mm window was required to cover the 95th error percentile. The surface model integrated information from the entire surface ROI to estimate marker displacement and resulted in the lowest errors. A 2.5 mm margin covered 95th error percentile for Traj1 and Traj4, and a 1-mm margin covered the 95th error percentile for the other trajectories.

Figure 7(d) shows a box plot comparison of the errors associated with the 4DCT+VRT model, and the 4DCT derived motion traces for m_{int5} displacements. These errors are reported in mm and were higher for trajectories with larger internal marker displacement. For the expanded 4DCT displacements, largest errors were observed for Traj2 and Traj3 and required an 8 and 6 mm window to cover the 95th error percentile [Fig. 7(e) and 7(f)]. For each marker, the 4DCT measurements reported a single-cycle representation that corresponded to the anatomy observed over the related duration of the 4DCT scan. While for all four trajectories, the single-cycle 4DCT representation was inadequate to capture the cycle-to-cycle variations and baseline shifts (e.g., Traj2), the magnitude of errors for Traj2 and Traj3 were higher due to the larger internal marker displacements associated with these two trajectories.

Integration of surface information with the 4DCT-derived centroids was found to produce a better estimation of marker motion over multiple cycles and resulted in smaller errors. The errors associated with the VRT + 4DCT are shown in box plot format in Fig. 7(d) and were on the order of slice thickness (3 mm) which presented a notable improvement to the 4DCT data alone. For all trajectories, largest errors appeared at points of peak inspiration and expiration where

FIG. 7. Comparing the performance of the models with 4DCT and FL. (a) Box plot of errors for estimated marker motion using the surface motion model (blue), the height + velocity motion model (green) and the scalar motion model (red). The scalar model did not take into account non-linear nature of the hysteresis and resulted in largest errors with the Sin_{OUT}. Errors for Traj1 and Traj4 were due to a change in correlation of AP and SI components of input trajectories. (b) Correlation of AP and SI components of Traj1. This trajectory exhibited a change in correlation at 154 s and at 242 s. (c) Correlation of AP and SI components of Traj4. A change in correlation after 110 s was observed. (d) Box plot comparison between model-estimated displacement errors of m_{int5} and 4DCT expansion. (e) Trajectories of m_{int5} motion exported from the model (blue), 4DCT (red), and ground truth FL (black) for Traj2. (f) Trajectories of m_{int5} motion exported from the model (blue), 4DCT (red), and ground truth FL (black) for Traj3. [Color figure can be viewed at wileyonlinelibrary.com]

the VRT + 4DCT model relied on extrapolation from the training data set.

4. DISCUSSION

In this work, we described a two-actuator externally and internally deformable lung phantom design which was a significant improvement over our previous design. Our previous phantom design used a single actuator to create deformable motion.²⁹ As a result, the phantom primarily moved in the SI direction and generated marker displacements that were linearly correlated with each other (little hysteresis from

response of the foam was present). Using that phantom, the curves that appear in Fig. 5(a)–5(c) only appeared as lines whose observed correlations were trajectory independent. In contrast to most commercial and investigational lung phantoms (including our previous single-actuator phantom), the current two-actuator phantom design allows the simulation of variable correlation between various parts of the thoracic anatomy. In our opinion, this capability is a major innovation that brings this phantom significantly closer to emulating human respiration. Human patients often exhibit such variable external–internal and internal–internal correlation because of differential engagement of thoracic and abdominal muscles from cycle to cycle and day to day. By ignoring this effect, current lung motion phantoms oversimplify the complexity of respiratory motion by assuming a constant correlation between surrogates and the tumor target. By simulating reproducibly variable correlation, our phantom design serves as an enabling tool for the development of realistic motion management techniques rather than the current “idealistic” motion management techniques based on simplified assumptions.

In these studies, our experimental setup was reproducible to be within ± 1 mm and the marker motions covered a range from a few mm to a few cm. Marker motions were generated for each programmed input trajectories and exhibited cycle-to-cycle variations and shifts that were based on the input. According to their distance from the actuators, each marker experienced a combination of the SI and AP components of the input trajectories. This produced deformable motion with different hysteresis observed for different points inside the phantom. Deformable motion and different hysteresis at different anatomical sites, with the mentioned displacement ranges are also characteristics of respiration-induced anatomical deformations in the human thorax.^{5,19} These characteristics are absent in the majority of currently available lung phantoms.^{1,22,23,27,29}

The deformable nature of the phantom’s exterior and foam insert result in deformable external–internal and internal–internal motion. Based on the input trajectory, the correlation between the internal–internal or internal–external markers can change. Parameters affecting this correlation include the amplitude of AP and SI input motion traces, their difference in phase, as well as the starting position for each actuator selected during the setup. Variations in these parameters were present in the patient-derived motion traces used in this study and produced corresponding changes in marker motion correlations. These characteristics mimic some of complexities that are present in respiration-induced motions of the human thorax and must be addressed when using more advanced motion management strategies such as real-time MLC tracking^{35,36} or respiratory gated delivery.^{37,38}

To date, the majority of clinically available tracking methods use an external surrogate such as markers on the surface and radiographic images^{39–41} to estimate the position of fiducials implanted within the tumor. A second group of methods, still under development, use an external surrogate and mathematical,^{15,16} or biomechanical¹⁸ models to generate

images of the underlying internal anatomy. To train the model, both groups of methods require an *a priori* observed correlation between the external surrogate and the internal anatomy. Examples of such correlations include the relation between 4DCT and the binning surrogate during initial simulation^{14–16,18} or concurrent kV images with surface monitoring during RT delivery.^{39,42} It has been well established that such external–internal correlation is prone to intra- and inter-fractional variations due to changes in the respiratory pattern, and anatomical shifts.¹¹ Our use of two independently programmable motion actuators enables the simulation of external–internal, and internal–internal motion correlations which depend on the input trajectory and, as observed in real-life, can change over time. Additionally, the external surface of the phantom is representative of its interior and can be used to estimate each marker's displacement. These qualities make our phantom a valuable tool to evaluate the performance of surrogate-based, image-guided motion management strategies.

As an application of our phantom, we evaluated the accuracy of different motion models constructed from images acquired via 4DCT or kV-fluoroscopy to estimate internal target positions. 4DCT is the current standard-of-care for thoracic RT, and is included as part of the initial simulation, and treatment planning process.¹ The volumetric motion models constructed from 4DCT images have the advantage of providing volumetric information regarding motion induced anatomical deformations.^{15,16,18} If properly validated in an in-situ setting, such models can have a dramatic impact in thoracic RT; some of their envisaged applications include adaptive real-time planning and delivery, as well as post-treatment verification of the delivered dose. However, the limiting factor associated with such models is their construction from *a priori* observed correlation between the external surrogate and the internal anatomy. Since this correlation is subject to change, these models are required to be updated with additional information relevant to the RT delivery session, which currently hinders their use in the clinic. kV images are commonly encountered in RT and used for patient simulation, motion assessment, and setup. The values of motion models constructed from kV images are the wide availability of this modality during RT delivery sessions and the potential to provide real-time information of the internal anatomy in the form of planar projections.⁴³ Limitations of 4DCT and kV fluoro are the associated imaging dose, as well as the convoluted and limited information present in kV radiographs. Motion models constructed from these images can greatly reduce the need for frequent kV imaging and the corresponding imaging dose while maintaining the ability to update the model over the duration of the treatment session. This approach is currently used in the SynchronyTM Respiratory Tracking System of the CyberKnife Robotic Radiosurgery System which enables real-time tracking of moving targets during RT delivery.¹⁹ A potential application of kV based motion models, which is currently studied in our group, is the

possibility of updating the aforementioned CT-based volumetric models using kV images. A methodology for such a model has been outlined by Li et al. and presents a virtual study generating volumetric CBCT images based on single kV images.⁴⁴ The compatibility of our phantom with these two imaging modalities, and its variable internal–external correlation makes it a valuable tool for the development and the quality assurance of such motion models.

It is important to note that several effects contribute to the multidimensional nature of respiratory-induced motion, such as engagement of several muscle groups with varying levels of involvement per cycle, the effect of cardiac and esophageal motion, and sliding motion between the lung lobes or sliding motion between the lung and chest wall. In lung cancer patients, the presence of additional complications, such as chronic obstructive pulmonary disease, emphysema, airway obstruction, embolisms, hemorrhage, or other lung injuries, in addition to the presence of tumors, produces additional factors that impact intra-thoracic motion. Although our proposed phantom can produce motions and deformations that are very close to human respiration, it cannot replicate the entire complexity of human respiratory motion. In addition, the programmed trajectory played in the phantom is the trace acquired from a point inside the lung (i.e., the location of the tumor), whereas it is applied along the spine and diaphragm in the phantom. Other limitations include the fact that in humans the expansion of the thoracic cavity is associated with the diaphragm, drawing inferiorly, whereas in the phantom, this directionality is reversed (i.e., an “inhalation” occurs in the phantom when the diaphragm moves superiorly).

In the present study, we embedded glass marbles and tungsten balls to study the motion at each location in the foam. Using deformable objects with different shapes instead of rigid spheres may allow us to study the effect of deformation of the tumor as a consequence of the motion on the amount of delivered dose to the planned target volume. Future work will also involve studying the effect of hardness of the tumor material on its deformation by testing objects with different durometers.

5. CONCLUSION

We presented the design and initial characterization of a programmable externally and internally deformable lung motion phantom. Our novel two-axis design enables the phantom to closely emulate several phenomena observed in human respiratory motion such as correlated external–internal and internal–internal motion between various parts of the thoracic anatomy, and variations in these correlations due to varying levels of engagement from thoracic and abdominal muscles, leading to variable hysteresis. We anticipate that such tools will become increasingly important for quality assurance of motion management technologies and techniques as our field moves toward increasingly potent forms of lung radiotherapy.

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